(54) Title: METHOD AND APPARATUS FOR REMOVING MOTION ARTIFACT AND NOISE FROM PULSE OXIMETRY

(57) Abstract

Motion compensation is based on analysis of intensity signals received by detectors, without separately measuring a motion signal, without providing feedback to cancel the motion signal and without attempting to mathematically eliminate the motion signal. Instead, the present invention mathematically recognizes the presence of the motion signal and recognizes a few key characteristics of the motion signal and makes corresponding assumptions. First, it is recognized that the motion looks in each wavelength signal is proportional. Second, it is assumed that the blood pulse signal is not affected by motion.

METHOD AND APPARATUS FOR REMOVING
MOTION ARTIFACT AND NOISE FROM PULSE OXIMETRY

BACKGROUND

The present invention relates to a pulse oximeter for detecting blood oxygenation, and in particular to the elimination of motion artifact which may affect the detected blood oxygenation signal.

Pulse oximeters typically measure and display various blood flow characteristics including but not limited to blood oxygen saturation of hemoglobin in arterial blood, volume of individual blood pulsations and the rate of blood pulsations corresponding to each heartbeat of the patient. The oximeters pass light through human or animal body tissue where blood perfuses the tissue such as a finger, an ear, the nasal septum or the scalp, and photoelectrically sense the change in absorption of light in the tissue. The amount of light absorbed is then used to calculate the amount of blood constituent being measured.

The light passed through the tissue is selected to be of one or more wavelengths that is absorbed by the blood in an amount representative of the amount of the blood constituent present in the blood. The amount of transmitted light passed through the tissue will vary in accordance with the changing amount of blood constituent in the tissue and the related light absorption.

The optical signal can be degraded by both noise and motion artifact. One source of noise is ambient light which reaches the light detector. Another source of noise would be electromagnetic coupling from other electronic instruments in the area. Motion of the patient can also affect the signal. For instance, when moving, the coupling between the detector and the skin or the emitter and the skin can be affected, such as by the detector moving away from the skin temporarily, for instance. In addition, since blood is a fluid, it may not move at the same speed as the surrounding tissue, thus

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resulting in a momentary change in volume at the point the oximeter probe is attached.

Such motion can degrade the signal used for making medical decisions, with the clinician being unaware of it.
5   This is especially true if there is remote monitoring of the patient, the motion is too small to be observed, the clinician is watching the instrument or other parts of the patient and not the sensor site, or in a fetus where motion is hidden.

In one oximeter system described in U.S. Patent No.
10 5,025,791, an accelerometer is used to detect motion. When motion is detected, readings influenced by motion are either eliminated or indicated as being corrupted. In a typical oximeter, measurements taken at the peaks and valleys of the blood pulse signal are used to calculate the desired
15 characteristic. Motion can cause a false peak, resulting in a measurement having an inaccurate value and one which is recorded at the wrong time. In U.S. Patent No. 4,802,486, assigned to Nellcor, the disclosure of which is incorporated herein by reference, an EKG signal is monitored and correlated
20 to the oximeter reading to provide synchronization to limit the effect of noise and motion artifact pulses on the oximeter readings. This reduces the chances of the oximeter locking on to a periodic motion signal. Still other systems, such as that set forth in U.S. Patent No. 5,078,136, assigned to
25 Nellcor, the disclosure of which is incorporated herein by reference, use signal processing in an attempt to limit the effect of noise and motion artifact. The '136 patent, for instance, uses linear interpolation and rate of change techniques to analyze the oximeter signal.
30   The nature of oximetry readings impose a number of difficulties in dealing with noise. The oximeter relies on mathematical analysis of the readings at two different wavelengths. Because different amounts of light are absorbed at each wavelength, the magnitude of the motion artifact due
35 to the same motion will be different for each signal. This is complicated by the fact that the lights are alternately

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pulsed, and thus each is influenced by a different amount of
motion, since the motion varies with time.

One system, set forth in PCT Publication No.
WO 92/15955 (Vital Signals, Inc.) correlates the non-noise
portion of two wavelength signals and generates a noise
reference signal. The noise reference signal is then provided
to an adaptive noise canceler to eliminate the noise from the
desired signal.

Patent No. 4,714,341 discloses the use of three
different wavelengths, rather than two, in order to detect
when noise is present. This patent teaches using the first
and second wavelength signals to produce a first oxygen
saturation value, and then using the first and third
wavelength signals to produce a second oxygen saturation
value. The two calculated values are then compared. If the
values are equal, as they should be absent motion, the signal
is presumed to be good. If the values are different, the
signal is assumed to contain motion and is disregarded.

SUMMARY OF THE INVENTION

The present invention is based on analysis of the
signal intensity received by the detectors, without separately
measuring the motion signal, without providing feedback to
cancel the motion signal and without attempting to
mathematically eliminate the motion signal. Instead, the
present invention mathematically recognizes the presence of
the motion signal and recognizes a few key characteristics of
the motion signal. First, although the magnitude of the
effect of motion on the signal intensity for each wavelength
will be different, the change in the logarithm of the motion
component will be approximately the same (for signals at
approximately the same time). This allows the motion
component to be cancelled out in a ratiometric equation.
Second, it is assumed that the blood pulse signal is not
affected by motion. This second assumption is more of an
approximation, since the blood pulse signal is somewhat

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affected by motion, which can actually change the blood volume
characteristics at any point in the patient.

5   The invention recognizes that the intensity signal
for each of the wavelengths includes a time-varying motion
term, and that this time-varying motion term is proportional
for each of the wavelengths. In addition, each wavelength
signal occurs close enough in time that the motion should not
vary noticeably, and can assumed to be the same for each
10   signal. Given this recognition, it is possible to determine
the saturation by including an appropriate time-varying motion
term in the equations to determine saturation. This can be
done for either a two wavelength or a three wavelength
embodiment.

15   In one two-wavelength embodiment, a time-variable
motion term corresponding to motion noise is included in the
equations representing the intensity for the first and second
wavelength signals. The logarithm of each equation is taken,
and then differentiated. The equations are then solved to
20   determine the saturation value by assuming that the motion is
a time varying function that is assumed to be independent of
the concentration, and not vary in the time between signals.

        In an alternate embodiment, radiation of three
discrete, different wavelengths is directed through a portion
25   of a patient. The amount of the radiation exiting the patient
is detected for each of the three wavelengths, producing three
intensity signals. Each intensity signal is represented by an
equation which is a function of a saturation, the-wavelength
corresponding to the intensity signal and corresponding
30   coefficients. In addition, a motion term is added to the
equation which is assumed to be variable with time and is
assumed to be the same for each of the different wavelength
intensity signals. The three equations are then solved to
determine the saturation value, preferably using matrix
35   algebra.

        For a fuller understanding of the nature and
advantages of the invention, reference should be made to the

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ensuing detailed description taken in conjunction with the
accompanying drawings.

BRIEF DESCRIPTION OF THE DRAWINGS

5   Figs. 1A-1D are diagrams of an intensity signal
showing the effects of pulsatile flow and motion noise;
           Figs. 2A and 2B are diagrams illustrating the effect
of motion on the path length of emitted light, and thus on the
intensity of received light; and
10   Fig. 3 is a block diagram of a system according to
the present invention.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

           Figs. 1A-1D illustrate aspects of a pulse oximeter
15   signal which the present invention takes advantage of.
Fig. 1A shows the logarithm of a detected infrared signal.
Fig. 1B shows the logarithm of a detected red wavelength
signal. For both of these figures, the signal includes motion
occurring in the interval of 5-12 seconds. Otherwise, both
20   the red and infrared signals are noise-free optical signals.
Fig. 1C shows the result of a subtraction between the signals
in Figs. 1A and 1B. As this illustrates, the subtraction
cancels out the noise. This is because the data exists in
logarithm form, and the motion corruption is additive.
25   Accordingly, in addition to calculating saturation, the
difference waveform (Fig. 1C) can be scaled, and then
subtracted from either the logarithm of the IR or red signal
to obtain an estimate of the motion noise. Fig. 1D shows this
estimate.
30   Fig. 2A illustrates one possible example of how
motion can effect the intensity signal. A light emitter 16 is
shown emitting rays 18 through a patient's finger 20. This is
detected by a detector 22. As can be seen, the distance from
the emitter to the detector, D, will determine the amount of
35   light emitted by the emitter reaching the detector, since
there will be a natural spreading effect of non-collimated

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light rays. The farther away the detector is, the more
spreading results.

Fig. 2B illustrates another example showing how
motion of a finger can compress and widen the finger
(exaggerated in the figure) and temporarily cause the light
emitter 16 to move away from the detector an additional
distance indicated by arrow 24. This additional distance will
cause less of the light to reach the detector, since there
will be more spreading of the light emitted at this larger
distance. This will result in a lower intensity waveform
being detected by the detector. Alternately, compression
could result in a higher intensity waveform. Motion and noise
can take other forms as well, and can vary for other reasons
than non-collimated light rays. For instance, the emitter and
detector could be slightly misaligned.

The present invention recognizes that the
calculation for determining oxygen saturation by pulse
oximetry using the "ratio of ratios" can be assumed to have a
motion term which is independent of any particular wavelength.
An understanding of this first requires an understanding of
how the ratio of ratios is calculated.

Using Lambert-Beer's law as a starting point,
equation (1) below is used to determine saturation in pulse
oximetry:

$$I(\lambda, t) = I_o(\lambda) \exp(-(s\beta_o(\lambda) + (1-s)\beta_r(\lambda))l(t)) \quad (1)$$

where: λ = wavelength
           t = time
           I_o = intensity of light transmitted
           I = intensity of light detected
           s = oxygen saturation
           β_o, β_r = empirically derived absorption coefficients
                    for oxygenated and deoxygenated hemoglobin,
                    respectively
           $l(t)$ = a combination of concentration and path
                    length from emitter to detector as
                    a function of time

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The traditional approach is to solve for ratio of ratios and then calculate saturation.

Take natural logarithm of equation (1) for IR and Red:

$$\log I = \log I_o - (s\beta_o + (1-s)\,\beta_r)\,I \qquad (2)$$

Differentiate equation (2) with respect to time:

$$\frac{d\log I}{dt} = -(s\beta_o + (1-s)\,\beta_r)\,\frac{dI}{dt} \qquad (3)$$

Divide Red (3) by IR (3)

$$\frac{d\log I(\lambda_R)/dt}{d\log I(\lambda_{IR})/dt} = \frac{s\beta_o(\lambda_R) + (1-s)\,\beta_r(\lambda_R)}{s\beta_o(\lambda_{IR}) + (1-s)\,\beta_r(\lambda_{IR})} \qquad (4)$$

For a discrete time sample, equations of the above form can be rewritten by noting:

$$\frac{d\log I(\lambda,t)}{dt} = \log I(t_2,\lambda) - \log I(t_1,\lambda)$$

Using $\log A - \log B = \log A/B$, the above equation can then be written as:

$$\frac{d\log I(\lambda)}{dt} = \log\!\left(\frac{I(t_2,\lambda)}{I(t_1,\lambda)}\right)$$

So,

$$\frac{\dfrac{d\log I(\lambda_R)}{dt}}{\dfrac{d\log I(\lambda_{IR})}{dt}} = \frac{\log\!\left(\dfrac{I(t_1,\lambda_R)}{I(t_2,\lambda_R)}\right)}{\log\!\left(\dfrac{I(t_1,\lambda_{IR})}{I(t_2,\lambda_{IR})}\right)} = R \qquad (5)$$

Where R is the "ratio of ratios."

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Solving (4) for s using (5) gives:

$$s = \frac{\beta_r(\lambda_R) - R\beta_r(\lambda_{IR})}{R(\beta_o(\lambda_{IR}) - \beta_r(\lambda_{IR})) - \beta_o(\lambda_R) + \beta_r(\lambda_R)}$$

From (5) note R can be calculated using two points corresponding to measurements at two different times, t.

5   Alternately, a family of points can be used.
To see this latter point define:

$$x(t) = \log\left(\frac{I(t+\Delta t, \lambda_{IR})}{I(t, \lambda_{IR})}\right)$$

$$y(t) = \log\left(\frac{I(t+\Delta t, \lambda_R)}{I(t, \lambda_R)}\right)$$

Then, equation (5) can be written as:

$$y(t) = Rx(t)$$

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and for a family of points over time this will describe a cluster of points that define a best-fit line of y versus x with a slope given by R.

The present invention modifies the above equations
15   by recognising that a term can be added to account for motion and noise. In particular, the motion and noise component can be represented by a function which varies with time and is wavelength-independent. This recognition allows a mathematical solution to isolate and eliminate the motion and
20   noise components without requiring prior art methods such as separately measuring the motion.

Motion. For example, to account for motion and noise, we can modify equation (1) by multiplying by a time varying function $\eta(t)$ representing wavelength-independent
25   motion or noise. This gives the following equation:

$$I(\lambda, t) = I_o(\lambda)\, \eta(t) \exp\left(-(s\beta_o(\lambda) + (1-s)\beta_r(\lambda))\, l(t)\right) \qquad (6)$$

We can then solve for s using the same steps as used above.

First, we take the logarithm:

$$\log I = \log I_o + \log \eta - (s\beta_o + (1-s)\beta_r) I$$

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Next, we differentiate with respect to time:

$$\frac{d\log I}{dt} = \frac{d\log \eta}{dt} - (s\beta_o + (1-s)\beta_r) \frac{dI}{dt} \qquad (7)$$

Then, we determine the ratio of Red to IR:

$$\frac{d\log I(\lambda_R)/dt}{d\log I(\lambda_{IR})/dt} = \frac{d\log \eta/dt - (s\beta_o(\lambda_R) + (1-s)\beta_r(\lambda_R)) \frac{dI}{dt}}{d\log \eta/dt - (s\beta_o(\lambda_{IR}) + (1-s)\beta_r(\lambda_{IR})) \frac{dI}{dt}}$$

10   Now if $d\log \eta/dt$ is large compare to the other terms the
ratio of ratios will be driven towards unity, driving s
towards a wavelength-dependant constant. So because in this
model optical coupling due to motion appears identically in
both wavelengths, its presence drives the saturation to this
15   wavelength-dependant constant.

The present invention thus allows a calculation of
blood oxygen saturation by mathematically recognizing the
motion signal. This enables a solution which does not require
separately measuring the motion signal, providing feedback to
20   cancel the motion signal, or attempting to mathematically
eliminate the motion signal. Set forth below are two
preferred embodiments for implementing the present invention,
one using three wavelengths of light and another using two
wavelengths.

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A Three-wavelength Solution

Let λ_o be some other wavelength (not IR or Red).
Now take the logarithm and differentiate this third
wavelength, obtaining (7). One approach might be to

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difference IR with this new wavelength, and similarly with Red. The problem with differencing is that R could become infinite when:

$$\frac{d}{dt} \log I(\lambda_{IR}) - \frac{d}{dt} \log I(\lambda_o) = 0.$$

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Here is a better solution. Rewrite (7) as:

$$\frac{d \log I}{dt} = \frac{d \log \eta}{dt} + (\beta_r - \beta_o) s \frac{dl}{dt} - \beta_r \frac{dl}{dt}$$

Now to introduce some matrix algebra, define:

$$b_i = \frac{d}{dt} \log I(\lambda_i)$$

$$u = \frac{dl}{dt}$$

$$m = \frac{d}{dt} \log \eta$$

$$c_{1j} = \beta_r(\lambda_j) - \beta_o(\lambda_j)$$

$$c_{2j} = \beta_r(\lambda_j)$$

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With this notation

$$\begin{bmatrix} b_1 \\ b_2 \\ b_3 \end{bmatrix} = \begin{bmatrix} c_{11} & c_{21} & 1 \\ c_{12} & c_{22} & 1 \\ c_{13} & c_{23} & 1 \end{bmatrix} \begin{bmatrix} su \\ u \\ m \end{bmatrix}$$

$$b = Cx$$
$$x = C^{-1}b$$
$$s = x_1/x_2$$
$$m = x_3$$

So as long as C is full rank, there is no difficulty in solving for saturation and the optical coupling terms
5   uniquely. In other words, you can now solve for m exactly because there is no wavelength where $b_1 = b2 = b_3$ for a given saturation.

Note a calibration set of extinction coefficients are needed for this third wavelength, but also note that the
10   best new wavelength is one that gives the highest condition number to C, which is not necessarily the isobestic point. The calibration coefficients for the third wavelength are constrained by the coefficients for the first two wavelengths. When there is no motion, the saturation calculated using two
15   wavelengths and three wavelengths should be the same.

This optical coupling method will be less accurate when the lumped concentration path-length term becomes wavelength dependent, then the dependence no longer ratios. away in calculating saturation. Also, there is no reason to
20   believe that u will look anything like a typical pulse oximetry waveform during motion since path-length and concentration will be varying with the motion, and these effects will be seen in u, but s will still be the correct saturation.
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A Two-wavelengths Solution

- With two wavelengths we have:

$$\begin{bmatrix} b_1 \\ b_2 \end{bmatrix} = \begin{bmatrix} c_{11} & c_{21} & 1 \\ c_{12} & c_{22} & 1 \end{bmatrix} \begin{bmatrix} su \\ u \\ m \end{bmatrix} \tag{8}$$

5   Two equations and three unknowns. One approach is
to return to calculating R by rewriting (8):

$$b_1 = v + m$$
$$b_2 = Rv + m$$

10 where m is the motion term, as defined earlier, R is the ratio
of ratios, and v is the signal with no motion.
        There are two key assumptions which make the
solution possible. First, although the magnitude of the
effect of motion on each intensity signal will be different,
15 the change in the logarithm of the motion component at two
different times will be the same (which assumes the different
time signal samples are adjacent or close together in time).
This allows the motion component to be cancelled out in a
ratiometric equation. The second assumption is that the
20 motion does not cause any effect on the remainder of the
equation. There is some effect, since motion can change the
pulse flow characteristics of the blood, but this is typically
a small effect compared to the motion when there is
significant motion present. By assuming that the motion has
25 no effect on any elements of the concentration measurement, we
assume that v and m are not related.
        Another assumption is that the amount of motion is
the same at the time of both intensity signal measurements for
the two wavelengths. This is a reasonable assumption since
30 the typical motion signal varies at a rate of around 1 Hz,
while the light pulsing frequency is typically at a rate of
1200 Hz.

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Assuming v and m are independent over time, $(v,m)=0$ for some defined inner product. Substituting for v and m yields:

$$\left\langle \frac{b_2 - b_1}{R-1}, \frac{b_2 - Rb_2}{R-1} \right\rangle = 0$$

5   Solving for the R that solves this equality yields:

$$R = \frac{(b_2 - b_1, b_1)}{(b_2 - b_1, b_2)}$$

There are two problems with this approach.

When R approaches one, $b_2 - b_1$ approaches zero, and the above equation approaches zero divided by zero. This fact is
10   not in itself a total problem for when $b_2 - b_1$ does approach zero you simply use $R=1$.

A more limiting problem is the assumption that $(v,m)=0$. Certainly the motion signal is independent of the arterial pulsatile signal, but during motion, v also has path-
15   length concentration effects in it that are highly correlated with m, thus biasing R away from its true value.

Fig. 3 is a block diagram of one embodiment of a pulse oximeter implementing the present invention. Light from LEDs 14 passes into patient tissue 18, and after being
20   transmitted through or reflected from tissue 18, the light is received by photosensor 16. Either two or three LEDs can be used depending upon the embodiment of the present invention. Photosensor 16 converts the received energy into an electrical signal, which is then fed to input amplifier 20.
25   Light sources other than LEDs can be used. For example, lasers could used, or a white light source could be used with appropriate filters either at the transmitting or receiving ends.

Time Processing Unit (TPU) 48 sends control
30   signals 68 to the LED drive 32, to alternately activate the

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LEDs. Again, depending on the embodiment, the drive may
control two or three LEDs.

The signal received from input amplifier 20 is
passed through three different channels as shown in the
embodiment of Fig. 3, for three different wavelengths.
Alternately, two channels for two wavelengths could be used.
Each channel includes an analog switch 40, a low pass
filter 42, and an analog to digital (A/D) converter 38.
Control lines 69 from TPU 48 select the appropriate channel at
the time the corresponding LED 14 is being driven, in
synchronization. A queued serial module (QSM) 46 receives the
digital data from each of the channels. CPU 50 transfers the
data from QSM 46 into RAM 52 as QSM 46 periodically fills up.
In one embodiment, QSM 46, TPU 48, CPU 50 and RAM 52 are part
of one integrated circuit, such as a DMC68HC16 microcontroller
from Motorola.

The method of the present invention is practiced by
CPU 50 on the data in RAM 52 as received through the various
channels from photodetector 16. The signal from
photodetector 16 is the signal which originated from LEDs 14,
as reflected or transmitted by patient 18, and including
undesired noise artifact.

As will be understood by those of skill in the art,
the present invention can be embodied in other specific forms
without departing from the spirit or essential characteristics
thereof. For example, saturation could be determined using
different mathematical calculations, once it is recognized
that the motion term is a function of time that is independent
of wavelength and is approximately the same for two adjacent
in time signal samples at two different wavelengths. In one
example, the mathematical determination could be done by
dividing the two intensity equations to eliminate the motion
term. Although this would only eliminate the motion from one
wavelength equation, this could be done for alternate
wavelengths in alternate samples. In a three wavelength
embodiment, division of two separate pairs could be done to
eliminate the motion signal. Accordingly, the disclosure of

the preferred embodiment of the invention is intended to be
illustrative, but not limiting, of the scope of the invention
which is set forth in the following claims.

WHAT IS CLAIMED IS:

1. A method for measuring saturation of a blood
constituent in a patient comprising the steps of:
   irradiating said patient with electromagnetic
radiation of at least two discrete, different
wavelengths;
   sensing an intensity of said radiation for each
of said wavelengths after it passes through a portion of
said patient to produce first and second intensity
signals; and
   determining said saturation by manipulating
said first and second intensity signals with the
assumptions that
      i) an amount of motion is the same at the
   same time for each of said intensity signals, and
         ii) the motion components of said
   intensity signals are proportional to one another.

2. The method of claim 1 wherein said determining
step assumes that the derivative of the logarithms of the
motion components of said intensity signals are the same.

3. The method of claim 1 further comprising the
steps of:
   representing each of said intensity signals as
a function of said saturation, the wavelength
corresponding to the intensity signal, and a time-
variable motion term corresponding to motion noise, said
motion terms being proportional to one another for each
of said intensity signals;
   taking the logarithm of each representation of
said first and second intensity signals;
   differentiating each logarithm;
   equating the first differentiated logarithm of
the first intensity signal to v + m, where m is the
portion of the signal due to motion;

15   equating the second differentiated logarithm of
16   the second intensity signal to $Rv + m$, where R is a ratio
17   of first and second wavelength ratios, each wavelength
18   ratio being the logarithm of the ratio of the intensity
19   signal for the wavelength at first and second times;
20   expressing said representations as a matrix;
21   solving said matrix for R by assuming v and m
22   are independent for some defined inner product; and
23   determining said saturation from R.

1   4. The method of claim 3 further comprising the
2   step of displaying said saturation on a monitor.

1   5. The method of claim 4 further comprising the
2   step of activating an alarm if said saturation is less than a
3   predetermined amount for a predetermined period of time.

1   6. The method of claim 1 further comprising the
2   steps of:
3   irradiating said patient with electromagnetic
4   radiation of at least three discrete, different
5   wavelengths;
6   sensing the intensity of said radiation for
7   each of said wavelengths after it passes through a
8   portion of said patient to produce first, second and
9   third intensity signals;
10   representing each of said intensity signals as
11   a function of said saturation, the wavelength
12   corresponding to the intensity signal, and a time-
13   variable motion term corresponding to motion noise, said
14   motion terms being proportional to one another for each
15   of said intensity signals; and
16   solving the three functions to obtain a value
17   for said saturation.

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1   7. The method of claim 1 wherein said determining
2   step assumes that the derivative of the logarithm of the
3   motion component of each intensity signal is the same.

1   8. The method of claim 6 wherein each of said
2   functions includes a plurality of coefficients, and further
3   comprising the step of determining a set of coefficients for
4   said third intensity signal from a measurement in the absence
5   of motion noise and a determination of said saturation from
6   said first and second intensity signals.

1   9. The method of claim 6 further comprising the
2   steps of:
3   taking the logarithm of each representation of
4   said first, second and third intensity signals;
5   differentiating each logarithm;
6   putting the differentiated logarithms into a
7   matrix; and
8   solving said matrix for said saturation.

1   10. A method for measuring the saturation of a
2   blood constituent in a patient comprising the steps of:
3   irradiating said patient with electromagnetic
4   radiation of three discrete, different wavelengths;
5   sensing the intensity of said radiation for
6   each of said wavelengths after it passes through a
7   portion of said patient to produce first, second and
8   third intensity signals;
9   representing each of said intensity signals as
10   a function of said saturation, the wavelength
11   corresponding to the intensity signal, and a time-
12   variable motion term corresponding to motion noise, said
13   motion term being the same for each of said intensity
14   signals; and
15   solving the three functions to obtain a value
16   for said saturation.

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1   11. The method of claim 10 wherein each of said
2 functions includes a plurality of coefficients, and further
3 comprising the step of determining a set of coefficients for
4 said third intensity signal from a measurement in the absence
5 of motion noise and a determination of said saturation from
6 said first and second intensity signals.

1   12. The method of claim 10 further comprising the
2 steps of:
3   taking the logarithm of each representation of
4 said first, second and third intensity signals;
5   differentiating each logarithm;
6   putting the differentiated logarithms into a
7 matrix; and
8   solving said matrix for said saturation.

1   13. A method for measuring the saturation of a
2 blood constituent in a patient comprising the steps of:
3   irradiating said patient with electromagnetic
4 radiation of two discrete, different wavelengths;
5   sensing the intensity of said radiation for
6 each of said wavelengths after it passes through a
7 portion of said patient separately to produce first and
8 second intensity signals;
9   representing each of said intensity signals as
10 a function of said saturation, the wavelength
11 corresponding to the intensity signal, and a time-
12 variable motion term corresponding to motion noise, said
13 motion term being the same for each of said intensity
14 signals;
15   taking the logarithm of each representation of
16 said first and second intensity signals;
17   differentiating each logarithm;
18   equating the first differentiated logarithm of
19 the first intensity signal to v + n, where n is the
20 portion of the signal due to motion;

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21   equating the second differentiated logarithm of
22   the second intensity signal to Rv + m, where R is a ratio
23   of first and second wavelength ratios, each wavelength
24   ratio being the logarithm of the ratio of the intensity
25   signal for the wavelength at first and second times;
26   expressing said representations as a matrix;
27   solving said matrix for R by assuming v and m
28   are independent for some defined inner product; and
29   determining said saturation from R.

1   14. An apparatus for measuring the saturation of a
2   blood constituent in a patient comprising:
3   first and second emitters, said emitters
4   emitting radiation of first and second different
5   wavelengths;
6   a detector for sensing the intensity of said
7   light, said detector being mounted relative to said first
8   and second emitters so that said light is detected after
9   it passes through a portion of said patient;
10   a controller for alternately activating said
11   emitters so that said detector detects the different
12   wavelengths at different times to produce first and
13   second intensity signals; and
14   control means for determining said saturation
15   by manipulating said first and second intensity signals
16   with the assumptions that
17   i) an amount of motion is the same at the
18   same time for each of said intensity signals, and
19   ii) the motion components of said
20   intensity signals are proportional to one another.

1   15. The apparatus of claim 14 wherein said control
2   means further comprises means for assuming that the derivative
3   of the logarithm of the motion components are the same.

1   16. The apparatus of claim 14 wherein said
2 apparatus is a pulse oximeter and said control means further
3 comprises:

4   means for representing each of said intensity
5 signals as a function of said saturation, the wavelength
6 corresponding to the intensity signal, and a time-
7 variable motion term corresponding to motion noise, said
8 motion term being the same for each of said intensity
9 signals;

10   means for taking the logarithm of each
11 representation of said first and second intensity
12 signals;

13   means for differentiating each logarithm;
14   means for equating the first differentiated
15 logarithm of the first intensity signal to v + m, where m
16 is the portion of the signal due to motion;
17   means for equating the second differentiated
18 logarithm of the second intensity signal to Rv + m, where
19 R is a ratio of first and second wavelength ratios, each
20 wavelength ratio being the logarithm of the ratio of the
21 intensity signal for the wavelength at first and second
22 times;

23   means for expressing said representations as a
24 matrix;

25   means for solving said matrix for R by assuming
26 v and m are independent for some defined inner product;
27 and

28   means for determining said saturation from R.

1   17. The apparatus of claim 14 further comprising a
2 display, coupled to said control means, for displaying said
3 saturation.

1   18. The apparatus of claim 17 further comprising an
2 alarm coupled to said control means for indicating when said
3 saturation is less than a predetermined amount for a
4 predetermined period of time.

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19. The apparatus of claim 14 further comprising:
       a third emitter for irradiating said patient
with electromagnetic radiation of a third discrete,
different wavelength;
       said controller alternately activating said
first, second and third emitters so that said detector
produces first, second and third intensity signals;
       said control means including
       means for representing each of said intensity
signals as a function of said saturation, the
wavelength corresponding to the intensity signal,
and a time-variable motion term corresponding to
motion noise, said motion term being the same for
each of said intensity signals; and
            means for solving the three functions to
       obtain a value for said saturation.

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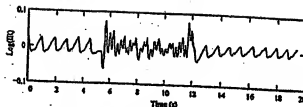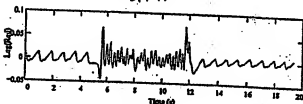
JA 29110

FIG. 1 (a)

FIG. 1 (b)
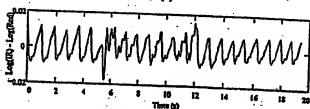
FIG. 1 (c)
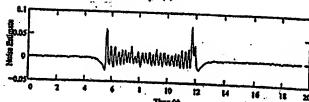
FIG. 1 (d)
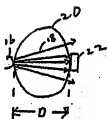
JA 29111

FIG. 2A
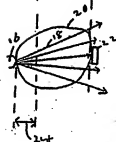
FIG. 2B
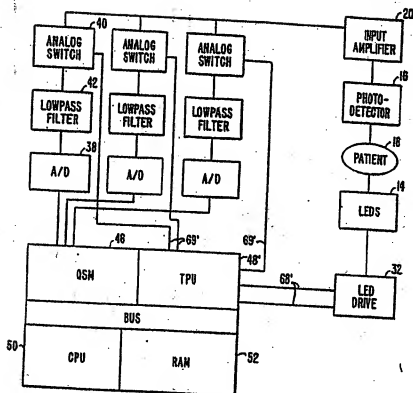
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FIG. 3

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INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER
IPC 6   A61B5/00

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
IPC 6   A61B   G01N   G06F

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category°	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	WO,A,94 03102 (UNIV SWANSEA ;PARKER DAWOOD (GB)) 17 February 1994 see the whole document	1,6,10
Y		2,7, 13-16,19

Y	US,A,5 351 685 (POTRATZ ROBERT S) 4 October 1994 see the whole document	2,7, 13-16,19
A		1,10,14

A	US,A,4 167 331 (NIELSEN LARRY L) 11 September 1979	1-3,6, 10, 12-14, 16,19
	see the whole document	

	-/--	

[X] Further documents are listed in the continuation of box C.   [X] Patent family members are listed in annex.

° Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier document but published on or after the international filing date

"L" document which may throw doubt on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
7 October 1996	-4. 11. 96

Name and mailing address of the ISA	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL - 2280 HV Rijswijk Tel. (+ 31-70) 340-2040, Tx. 31 651 epo nl, Fax: (+ 31-70) 340-3016	Ferrigno, A

Form PCT/ISA/210 (second sheet) (July 1992)

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C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	DE,A,36 29 447 (CRITICARE SYSTEMS INC) 9 April 1987 see the whole document	1,3-5, 10,13, 14,17,18

Form PCT/ISA/210 (continuation of second sheet) (July 1992)

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Box I   Observations where certain claims were found unsearchable (Continuation of Item 1 of first sheet)

This International Search Report has not been established in respect of certain claims under Article 17(2)(a) for the following reasons:

1. [X] Claims Nos.:   1-13
   because they relate to subject matter not required to be searched by this Authority, namely:
   .See additional sheet.

2. [ ] Claims Nos.:
   because they relate to parts of the International Application that do not comply with the prescribed requirements to such
   an extent that no meaningful International Search can be carried out, specifically:

3. [ ] Claims Nos.:
   because they are dependent claims and are not drafted in accordance with the second and third sentences of Rule 6.4(a).

Box II   Observations where unity of invention is lacking (Continuation of Item 2 of first sheet)

This International Searching Authority found multiple inventions in this international application, as follows:

1. [ ] As all required additional search fees were timely paid by the applicant, this International Search Report covers all
   searchable claims.

2. [ ] As all searchable claims could be searches without effort justifying an additional fee, this Authority did not invite payment
   of any additional fee.

3. [ ] As only some of the required additional search fees were timely paid by the applicant, this International Search Report
   covers only those claims for which fees were paid, specifically claims Nos.:

4. [ ] No required additional search fees were timely paid by the applicant. Consequently, this International Search Report is
   restricted to the invention first mentioned in the claims; it is covered by claims Nos.:

Remark on Protest   [ ] The additional search fees were accompanied by the applicant's protest.

                                     [ ] No protest accompanied the payment of additional search fees.

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FURTHER INFORMATION CONTINUED FROM   PCT/ISA/210

The subject-matter of claims 1-13 relates to a diagnostic
method carried out on the living human body. According to
Rule 39 and Article 17 PCT, no International Search is
required for such a subject-matter. An incomplete search
has been therefore carried out for claims 1-13: the search
has been limited to the "means" for carrying out the claimed
method. Claims 14-19 have been searched completely.

JA 29117

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
WO-A-9403102	17-02-94	AU-A- 4719893	03-03-94
		ZA-A- 9305579	02-02-94
US-A-5351685	04-10-94	US-A- 5533507	09-07-96
US-A-4167331	11-09-79	DE-A- 2756462	22-06-78
		JP-C- 1345150	29-10-86
		JP-A- 53068778	04-08-78
		JP-B- 61011097	01-04-86
DE-A-3629447	09-04-87	JP-A- 62109547	20-05-87

Form PCT/ISA/210 (patent family annex) (July 1992)